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(54) Thermographic apparatus for measuring the temperature distribution in a substantially dielectric medium.

(57) A thermographic apparatus is provided for measuring the temperature distribution in a substantially dielectric medium by detecting the microwave energy emerging from the medium. The apparatus is particularly useful for detecting carcinomas of the human body. The apparatus comprises an array of microwave antennas positioned adjacent the patient's body and electronic means for processing the broad band signals induced in the microwave antennas to determine the temperature prevailing in adjacent volume elements of the patient's body. The electronic means is laid out to process the signals induced in each of the antennas and to correlate each signal with the signal from one of the antennas in a fixed reference location. The correlation produces first and second values containing information relating to the amplitude and phase of the signal received in the or each detecting position of that antenna relative to the amplitude and phase of the signal received by the reference antenna. A computer is used to carry out an inverse transformation of the relative amplitude and phase signals from all the antennas to produce a matrix of temperature values relating to adjacent volumes of the patient's body.

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Thermographic Apparatus for Measuring the  
Temperature Distribution in a Substantially  
Dielectric Medium

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The present invention relates to thermographic apparatus for measuring the temperature distribution in a substantially dielectric medium by detecting the microwave energy emerging from the medium, and has particular reference to thermographic apparatus for use in medical applications. Such medical applications include the detection of carcinomas or lesions of the human body which manifest themselves by a locally elevated temperature and increase in emission of black-body radiation. In this case the human body forms the substantially dielectric medium. The thermographic apparatus can however also be used to determine temperature distributions in other dielectric media.

So far as the medical applications are concerned it is known that the region surrounding a carcinoma usually shows an excess of temperature over the mean temperature of the human body, and that this effect is extended to a larger region by the flow of blood from the excess of blood vessels found around the carcinomas. This means that one can in practice detect the carcinomas with a local temperature excess of only  $1.5^{\circ}$  K using thermographic

1 apparatus. As the temperature distribution is measured  
solely by detecting the microwave energy emerging from the  
patient's body the techniques of thermography are passive  
and non-invasive and can thus be used for the routine screen-  
5 ing of people for cancer or other infections without any  
danger of side effects. Existing thermographic apparatus  
has been successfully used for the detection of carcinomas,  
in particular carcinomas of the female breasts, it is not  
however really suitable for mass screening because of  
10 the large time required to take the necessary measurements.

Typical prior art thermographic apparatus and methods are  
described in the articles "Detection of Breast Cancer  
by Microwave Radiometry" by A.H. Barrett, P.C. Myers in  
15 the journal "Radio Science 12" (supplement) 1977, and  
"Centimeter- and Millimeter-Wave Thermography - A survey  
on Tumor Detection" by J. Edrich in "The Journal of  
Microwave Power" vol. 13, No. 2, 1979, pages 95 to 104.  
In the first mentioned article a rectangular waveguide  
20 antenna filled with a low loss solid with a dielectric  
constant equal to 11 is placed flush against the patient's  
skin and a plurality of readings are taken at different  
locations, typically nine different locations on each  
breast. In each position the antenna receives signals from  
25 a substantial volume of the patient's body and a relative-  
ly high reading indicates that a carcinoma may be present  
in that particular volume of the patient's body. The  
carcinoma can be more precisely localised by comparing the  
readings from adjacent positions of the antenna. It is  
30 possible, although not discussed by these authors, to use an  
array of independent antennas to give several simultaneous  
readings. This would shorten the time taken for an examin-  
ation, but would not give any increase in spatial resolution.  
In the method described in the second article the patient  
35 is laid on a table and a receiving arrangement, comprising  
a reflector and a horn antenna mounted in front of the  
reflector, is mounted above the patient and is movable

1 within a cartesian coordinate system. In this way the  
microwave radiation emerging from the patient's body is  
directed by the reflector to the horn antenna and different  
areas of the body can be scanned by moving the receiver  
5 arrangement to different coordinate positions. Again a  
plurality of readings have to be taken and this takes a  
substantial amount of time.

Another method of thermography is described in the article  
10 "New Correlation Radiometer for Microwave Thermography"  
by A. Mamouni, J.C. van de Velde and Y. Leroy in Electronic  
Letters August 6th, 1981, vol. 17, No. 16. This article  
describes a receiver which does not consist of a single  
probe (or antenna) as in the usual microwave radiometers,  
15 but consists of a combination of two or several probes  
adjacent to each other and in contact with the patient's  
body. The idea underlying this arrangement is that each  
probe will receive microwave signals from an adjacent  
volume of tissue and that two adjacent volumes of tissue  
20 associated with two adjacent probes will have a region of  
overlap common to both probes. The electronic circuitry  
associated with the two probes is said to be such that the  
contribution of the thermal noise generated in the common  
volume of overlap can be identified. It is stated that  
25 a delay introduced into one of the two signal paths from the  
two probes can be used to modify the contribution of the  
different sub-volumes in the volume of overlap. The problem  
with this arrangement however is that it is only intended  
to produce signals from the regions of overlap between ad-  
30 jacent probes which will be small, giving better resolution  
than with single probes, but temperature mapping of a whole  
area of a body is again only possible by producing relative  
movement between the probes and the body. Thus again a  
plurality of readings is necessary, which takes an un-  
35 desirably long time. Furthermore, the spatial resolution  
of this and other prior art thermographic apparatus are  
relatively restricted.

1 The principal object underlying the present invention is  
to provide thermographic apparatus which is able to  
give an image of a whole area of a dielectric medium, in  
particular of a patient's body, and to measure the tempera-  
5 ture distribution in the medium with substantially improved  
spatial and thermal resolution in a short time.

In order to satisfy this object there is provided, in  
accordance with the invention, and starting from the prior  
10 art apparatus described in the article in Electronics  
Letters as cited above, thermographic apparatus for  
measuring the temperature distribution in a substantially  
dielectric medium by detecting the microwave energy  
emerging from the medium, the apparatus comprising an  
15 array of microwave antennas positioned adjacent the dielec-  
tric medium and electronic means for processing the broad-  
band signals induced in the microwave antennas in a  
plurality of detecting positions to determine the temper-  
ature prevailing in a volume element of the dielectric  
20 medium, characterised in that the electronic means com-  
prises means for processing the signal induced in each  
of the antennas in each of the detecting positions and for  
correlating each signal with the signal from at least one  
of said antennas in a fixed reference location to produce,  
25 for the or each detecting position of each antenna, first  
and second values containing information relating to the  
amplitude and phase of the signal received in that de-  
tecting position relative to the amplitude and phase of the  
signal received by at least one of the reference antennas;  
30 and computing means for forming an inverse transformation of  
the first and second values associated with each antenna of  
the array, said inverse transformation being a matrix of  
temperature values relating to adjacent volumes of said  
dielectric medium.

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This apparatus is thus based on the principle of aperture  
synthesis as used in radio astronomy from which it is known

1 that the distribution of amplitude and phase over an antenna's aperture is the Fourier transform of the source's brightness distribution. This is a significant extension of correlation techniques to give an image of the brightness  
5 distribution in the whole of the region of response of the individual antennas, which are made so that a large region of the patient's body is in each and every antenna's response pattern. Using the apparatus of the invention it should be possible to measure the temperature distribution  
10 over a whole region of a patient's body of approximately the same size as the array size within a very short space of time and with high spatial and thermal resolution. The short period of time required for the measurement not only makes the apparatus suitable for mass screening techniques,  
15 but also for monitoring the effects of treatments to the patient on virtually any time scale. Since the signals are correlated across the extremities of the array, the apparatus will have the high resolution appropriate to the array size over the whole region examined. It is believed  
20 that the presently proposed apparatus will be capable of detecting deep-seated hot spots  $1.5^{\circ}$  K above ambient temperature with a positional accuracy of a few millimeters in a short time.

25 The inverse transformation of the first and second values is preferably an inverse Fourier transformation. The inverse Fourier transformation is however theoretically only valid for temperature sources which are at a substantial distance from the antennas. If a straightforward inverse Fourier  
30 transformation produces results which are obviously inconsistent it may be necessary to make certain corrections, or to resort to an inverse Fresnel transformation, or even to make corrections and use an inverse Fresnel transformation.

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In view of this; one embodiment of the apparatus is characterised by the provision of phase adjustment means

1 for adjusting the relative phase of each signal induced in  
each of the antennas in the or each detecting position,  
relative to the phase of the signal received by at least one  
of the reference antennas, prior to effecting the inverse  
5 transformation, whereby to compensate for near field  
effects. Thus this embodiment envisages that the corrections  
required to the inverse Fourier transformation, or possibly  
to the inverse Fresnel transformation can be made by correct-  
ing the relative phases of the signals prior to effecting  
10 the inverse transformation. This adjustment of the relative  
phases is most effectively made by software embodied in the  
computing means which will enable phase corrections to be  
effected and the output to be analysed. If at all possible  
it is preferable to use an inverse Fourier transformation  
15 rather than an inverse Fresnel transformation because the  
calculations involved in performing an inverse Fourier  
transformation are substantially easier.

A particularly preferred development of the apparatus is  
20 characterised in that an inert dielectric layer of high  
permittivity, preferably greater than 10, and of low loss,  
preferably better than 5 %, is placed between said array  
and said medium. The inert dielectric layer is preferably  
greater than 5 cm thick and should preferably be con-  
25 toured so that it at least approximately fits the shape  
of the patient's body, in order to avoid a skin/air inter-  
face which would result in substantial and undesirable  
losses. In this way the array of antennas can also be  
spaced further away from the patient without significant  
30 loss in signal strength. Because the array is spaced  
further from the patient the inverse transformation may be  
simplified so that it is straightforwardly possible to use  
an inverse Fourier transformation.

35 In a further development of the apparatus each of the  
antennas may be replaced by two or more antenna elements,  
whereby separate polarisation components can be measured

1 either prior to or after the inverse transformation, and means may be provided for compensating for polarisation differences.

5 This embodiment recognises that it may be essential to take account of the state of polarisation of the signals received by the individual antennas.

In a particularly preferred embodiment means are provided  
10 for varying the centre frequency and/or the bandwidth at which the measurement of the temperature distribution is carried out.

This embodiment is based on the recognition that the  
15 correct choice of frequency and/or bandwidth can yield more information on the temperature distribution inside the patient's body. In particular, the taking of several measurements at different centre frequencies may be necessary for a depth dimension to be added to the measured  
20 temperature distribution. It will be appreciated that the measured temperature distribution may be regarded as a two-dimensional projected temperature distribution, similar to an X-ray which is a two-dimensional projected density distribution. A theoretical discussion of the points to be  
25 considered when selecting frequency and bandwidth is included at the end of this specification.

In one practical embodiment for varying said centre frequency said electronic means comprises a local oscillat-  
30 or and a mixer for generating a difference frequency, with the local oscillator frequency being variable whereby to vary said centre frequency. This embodiment may include amplifier and/or filter means for varying said bandwidth.

35 In an alternative arrangement the electronic means does not include a local oscillator but is characterised in that the centre frequency and bandwidth are both determined by an amplifier and/or a filter.



1 In a further practical embodiment the array of microwave  
antennas is spatially fixed, so that each antenna has one  
detecting position, and switching means is provided for  
connecting said antennas groupwise in turn to said  
5 electronic means. In this way it is possible to provide  
a large number of antennas in the array without having to  
provide a separate electronic channel for each individual  
antenna, which would be very costly. Instead selected  
groups of antennas are connected in turn to a like number  
10 of electronic channels.

Alternatively, a smaller number of antennas is arranged in  
an array and means can be provided for moving the array  
so that each antenna, other than the reference antenna or  
15 antennas, adopts or moves through a plurality of detecting  
positions. In this embodiment a respective electronic  
channel is associated with each antenna, however the total  
number of antennas is substantially reduced.

20 The movable array can be a linear array, which may be  
rotated or moved laterally, and which is relatively simple  
to construct, it can however also comprise a two-dimension-  
al array of non-uniformly spaced antennas. This latter  
arrangement is particularly advantageous because the lay-  
25 out of the individual antennas in the array can be selected  
so that cross-talk between the individual antennas is avoid-  
ed.

Said plurality of detecting positions preferably fill the  
30 area over which said array is moved. This embodiment recog-  
nises that the amount of information which can be gathered  
is greatest if, in the course of taking the measurements,  
a measurement is made at each element of the area over  
which the said array is moved.

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In a further development of the apparatus intermediate  
storage means is provided for temporarily storing said

1 signals from each of said antennas in the or each detect-  
ing position. This embodiment makes it possible to tempo-  
rarily store the signals so that they may be subsequently  
fed to the computing means in the correct time sequence.

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In a particularly preferred embodiment the electronic means  
includes a noise calibration system with a noise generating  
source, such as a noise diode, adapted to feed noise to  
each of said antennas and said calibration system

10 is optionally adapted to calibrate the apparatus to take  
account of reflections at the surface of the medium and/or  
at boundaries within the medium.

The noise calibration system enables drift in the electron-  
15 ics to be recognised and makes it possible for suitable  
corrective measures to be taken. In a further development  
automatic gain control means are provided, being optionally  
controlled in dependence on a signal derived from the said  
noise calibration system.

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In a particularly preferred embodiment the electronic means  
comprises first multiplier means for forming the products  
of the signal from at least one of the reference antennas  
with the signal induced in each of the remaining antennas in  
25 the or each detecting position to yield said first values;  
phase shifting means for shifting the phase of the signal  
induced in each of the remaining antennas in the or each  
detecting position by  $90^\circ$  to produce phase shifted signals;  
and second multiplying means for forming the products of  
30 the signal from the same reference antenna with each of the  
phase shifted signals to yield said second values. When  
using this embodiment the first and second values have the  
general form of the product of an amplitude term and the  
sine and cosine of a phase difference term respectively.  
35 The phase and amplitude of the signal received in each an-  
tenna in the or each detecting position relative to the sig-  
nal received by the reference antenna can be directly  
calculated from the first and second values.

1 The antennas preferably comprise dipole antennas or antennas formed as printed circuits on dielectric material as all of the region in the patient being examined must be in the primary beam of each antenna at all times during the  
5 examination. Such antennas have the advantage of being both inexpensive and effective and a second, crossed dipole can be easily added to each dipole antenna for polarisation measurements.

10 Finally, the present invention recognises that existing hyperthermia apparatus, i.e. apparatus used to treat carcinomas by heating them to elevated temperatures, could be dramatically improved by combining it with the presently proposed thermographic apparatus. Such combined apparatus  
15 will be characterised by means for operating the thermographic apparatus and the hyperthermia apparatus in alternative time periods, optionally using the same antennas, whereby to detect the temperature distribution produced by the hyperthermia apparatus and, optionally, by means for  
20 controlling the hyperthermia apparatus using the signals from the thermographic apparatus.

Because the thermographic apparatus of the present invention should produce, for the first time, a very rapid  
25 read-out of the temperature distribution within the patient's body it should be possible to energise the hyperthermia apparatus for a short while, to measure the actual temperature distribution produced using the thermographic apparatus, to correct the relative phases and  
30 amplitudes of the microwave signals fed to the antennas of the hyperthermia apparatus, to carry out further heating to re-measure the temperature distribution, and to repeat this process so as to ultimately produce a desired temperature distribution in the body using the hyperthermia  
35 apparatus.

1 In other words the apparatus of the present invention  
could be used to provide a real time image of the temperat-  
ure distribution around a site being subjected to hyper-  
thermia treatment, with the image scanning process being  
5 time-shared with the application of hyperthermia power.  
With this arrangement it will of course be necessary to  
provide adequate thermal protection for the receiver  
preamplifier stages. If the time sharing intervals are  
small, say 0.1. seconds, the cooling effects of the bio-  
10 heat transport mechanisms around the area will be negli-  
gible, thus the thermogram represents the true temperature  
distribution during the hyperthermia process. This should  
greatly aid the preferential destruction of the cancers and  
the implementation of high resolution phased array hyper-  
15 thermia systems. The apparatus of the present invention  
should give a high enough resolution to match the size of  
area warmed by the hyperthermia treatment.

A further technique which might be useful is to shine a  
20 narrow band noise source thorough the body. The power  
required for an adequate signal is quite small (1 mW) and does  
not represent a health hazard. This signal would be blinked  
and detected synchronously in the computer software. The  
method may reveal variations in the opacity of the tissue  
25 as a function of frequency which can be detected by image  
comparison and may be found in clinical practice to  
correlate with abnormalities.

Embodiments of the invention will now be described in  
30 further detail by way of example only and with reference  
to the accompanying drawings which show:

Fig. 1 a table adapted to support a patient during a  
thermographic investigation,

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Fig. 2 a plan view of a fixed square antenna array,

Fig. 3a a plan view of rotatable linear antenna array,

1 Fig. 3b a plan view of a laterally movable linear antenna array,

Fig. 4 a schematic section on the line IV-IV of the array  
5 of Fig. 3a illustrating the position of a hot spot which is to be detected,

Fig. 5 a schematic block diagram of the arrangement of  
the electronic circuitry used to process the  
10 signals from the antennas of the array of Figs. 3 and 4,

Fig. 6 a schematic block diagram illustrating in more  
detail the construction of the correlation blocks  
15 shown in the block diagram of Fig. 5,

Fig. 7 a diagram illustrating the noise calibration  
principle underlying the circuitry of Figs. 5 and 6,  
and

20

Fig. 8 the temperature difference seen just into a dielectric for:

a) an isolated hot spot of diameter 1 cm with  
 $T = 1.5^{\circ} \text{K}$  above normal body temperature as seen  
25 through various thicknesses of muscles,

b) a layer of warmer material seen under 2 mm of  
skin and 1 cm of fat for various thicknesses of  
muscle for air ( $\epsilon_r = 1$ ) and dielectric  
( $\epsilon_r = 30$ ). In both cases the material has the  
30 electrical properties of muscle.

Turning firstly to Fig. 1 there can be seen a table 10  
supporting a slab of dielectric 11 of high relative  
permittivity  $\epsilon_r = 30$  and of low loss (less than 5%). A  
35 material of this kind is sold under the trademark "HiK" by  
Emerson and Cuming, Canton, Massachusetts 02021, USA. The  
upper surface of the slab of dielectric is provided with

1 moulded contours 12 which correspond approximately to the  
shape of the human body. Alternatively, slabs of moulded  
material could be slidably arranged on the basic slab of  
dielectric 11 so that they could be moved to provide a best  
5 fit for any particular patient. An antenna array 13 is  
positioned immediately beneath the slab of dielectric 11 in  
contact therewith and can be slid to and fro beneath the  
table on a pair of horizontal rails 14, 15, so that it can  
be positioned under the part of the patient's body under  
10 investigation.

The antenna array shown in Fig. 1 is in fact a rotatable  
linear antenna array as shown in Fig. 3a) and is rotated  
by a motor 16 disposed centrally beneath the array. The  
15 drive leads for the motor and the connections between the  
electronics mounted on the antenna array and the further  
electronics are schematically illustrated by the reference  
numeral 17. The antenna array may alternatively be a  
fixed antenna array such as the square antenna array 13'  
20 shown in Fig. 2.

In the square antenna array of Fig. 2 there are eight rows  
and eight columns of antennas 18, 19 each in the form of a dipole  
antenna with one of the antennas 18 forming a reference  
25 antenna. Although it would be theoretically possible to  
derive signals from each of said antennas simultaneously  
this would in fact require 64 processing channels which  
would probably be prohibitively expensive. For this reason  
it is preferred to provide say eight channels and to  
30 use electronic switching techniques to switch between  
selected groups of eight antennas. One of the antennas, the  
reference antenna 18, must of course always be connected  
to one of the channels, the reference channel, because the  
existence of a reference antenna is crucial to the meas-  
35 urement process as will later become clear from the dis-  
cussions of Figs. 5 and 6.

1 The groups of antennas which are connected to the eight  
channels at any time are preferably spread across the sur-  
face of the array so that cross-talk between individual  
antennas is minimised. This is illustrated in Fig. 2 which  
5 only shows antennas in eight distinct fields of the array,  
namely the antenna 18, which is the reference antenna, and  
the further antennas 19 which constitute one of the groups  
of antennas which are connected to the remaining seven  
electronic channels. It will be understood that identical  
10 antennas are provided in each field of the array, these  
antennas have however been omitted in order to simplify  
the illustration.

Rather than providing such a large number of individual  
15 antennas it is also possible to provide fewer antennas  
and to move the array to different detection positions as  
illustrated in Fig. 3a + b. In this arrangement it is how-  
ever essential that one antenna, in this case the antenna  
18, should be fixedly positioned for comparison purposes.  
20 The linear array of Fig. 3 is rotatable about the central  
axis 20 through  $180^{\circ}$  through a plurality of distinct  
detecting positions 21. This can be effected by the motor 16  
of Fig. 1 which can be either a stepping motor or a con-  
tinuously rotating motor. I.e. readings can be taken while  
25 the array is stationary or when it is moving.

An alternative arrangement is shown in Fig. 3b in which  
two linear arrays are positioned on either side of a central  
fixed reference antenna 18 and can be displaced linearly to  
30 the left and to the right relative to the central antenna  
as illustrated by the arrows V. This embodiment has the  
advantage that the antennas do not rotate and thus respond  
to the same polarisation components during the course of  
the examination.

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The array of Fig. 3a is shown again in cross-section in  
Fig. 4 which also shows the slab of dielectric 11 and a hot

1 spot 22 which is assumed to lie somewhere in the patient's  
body.

Fig. 4 also shows radiation emitted from one point in the  
5 hot spot which propagates in the general direction of the  
lines 23 and 24 and is detected by the two antennas 18  
and 19. It can easily be seen that the signal induced in  
the antenna 18 will have a different phase from that of the  
signal induced in the antenna 19. Of course the antenna 18  
10 and the antenna 19 are not just receiving radiation from the  
hot spot 22 but also from every other point in the patient's  
body such as the point 22'.

The antenna's beam pattern (its response to radiation from  
15 angles  $\theta$  off its principal axis) must be broad enough to  
ensure that it covers all the area to be examined. That  
is to say that the whole of the area to be examined should  
be in the primary beam of every antenna at all times.

20 It can be seen from Fig. 4 that the paths 23 and 24 will be  
mainly in the lossy medium of the patient's body when the  
hot spot 22 is at the edge of the area examined. This may  
degrade the signal-to-noise ratio so an optional solution  
is to take another antenna 18' and use that as a secondary  
25 reference antenna. The output from this antenna must also  
be combined with the output from the other antennas in the  
manner described below to give secondary desired first and  
second values. These secondary values can then, in principle,  
be eliminated during processing prior to the inverse  
30 transformation. The use of this technique does not change  
the principles underlying the following discussion.

The signals generated at the antennas 18 and 19 from the  
radiation emerging from the hot spot 22 may be described  
35 by formulae of the form:



1

$$V_{18} = V_{22} L_{23} f(\theta_{18}) \cos \left( \omega t - \frac{2\pi a}{\lambda} \right) \quad (A)$$

$$V_{19} = V_{22} L_{24} f(\theta_{19}) \cos \left( \omega t - \frac{2\pi b}{\lambda} \right) \quad (B)$$

where  $V_{22}$  is the signal leaving the point 22;  
 $L_{23}$  and  $L_{24}$  are the losses along the paths 23  
 and 24 of Fig. 4 respectively;  
 $f(\theta)$  is the beam pattern of the individual  
 antennas (assumed to be the same for all anten-  
 nas), with the angles  $\theta_{18}$ ,  $\theta_{19}$  being defined as  
 shown in Fig. 4; and  $\omega$  is the angular frequency  
 of the radiation leaving the point 22;  
 $a$  and  $b$  are the lengths of the paths 23 and 24  
 respectively, and  $\lambda$  is the wavelength in the  
 dielectric.

The expressions for the signals arising at the antennas 18  
 and 19 from the point 24 are similar vis:

$$V'_{18} = V'_{22} L_{23} f(\theta'_{18}) \cos \left( \omega t - \frac{2\pi a'}{\lambda} \right) \quad (C)$$

$$V'_{19} = V'_{22} L_{24} f(\theta'_{19}) \cos \left( \omega t - \frac{2\pi b'}{\lambda} \right) \quad (D)$$

and indeed similar equations can be written for the energy  
 arriving at the two antennas from each point in the  
 patient's body.

On summing these signals one obtains

$$\langle V_{18} \rangle = \iint V(\theta, a) \cdot L(\theta, a) \cdot f(\theta) \cdot \cos \left( \omega t - \frac{2\pi a}{\lambda} \right) \cdot d\theta da \quad (E)$$

$$\langle V_{19} \rangle = \iint V(\theta, b) \cdot L(\theta, b) \cdot f(\theta) \cdot \cos \left( \omega t - \frac{2\pi b}{\lambda} \right) \cdot d\theta db \quad (F)$$

- 1 These can be expressed in the form  $\langle V_{18} \rangle = \tilde{V}_{18} \cos$   
     $(\omega t - \psi_{18})$  and  $\langle V_{19} \rangle = \tilde{V}_{19} \cos(\omega t - \psi_{19})$  where  $\tilde{V}_{18}$  and  
     $\tilde{V}_{19}$  are mean voltages and  $\psi_{18}$  and  $\psi_{19}$  are mean phase  
    shifts and these signals now have to be processed and  
5 correlated to obtain the amplitude and phase of the net  
    signal received at antenna 19 relative to the signal received  
    at antenna 18. This processing is done by the electronic  
    circuitry shown in Figs. 5 and 6.
- 10 Referring firstly to Fig. 5 it can be seen that the signals  
    received by each of the antennas is directed through a  
    respective channel. Each channel first contains a circulator  
    or 25 which is a device for ensuring that energy flows  
    primarily from the antenna to the detection channel and  
15 not vice versa. Each circulator 25 is followed by a band-  
    pass filter 26 which may be incorporated in the preamplifiers  
    27.

- Thereafter the signals in each channel are amplified in  
20 preamplifiers 27 and are passed to image rejecting circuits  
    28 which reduce the noise present in the mixer image band,  
    coming from the broadband preamplifiers. These filters  
    reduce the uncorrelated noise present in the signals thus  
    improving the signal-to-noise ratio. At this stage the  
25 signals are fed into respective single-sideband mixers 29  
    and are mixed with a signal supplied by a local oscillator  
    30, which is common to all electronic channels. The  
    difference frequencies produced by the mixers 29 are then  
    fed to low pass filters 29' and then to the respective  
30 frequency amplifiers 31. The signal in the reference  
    channel associated with the reference antenna 18 is then  
    passed through a phase switching device 32 which will be  
    described later. Thereafter the signal from the reference  
    channel is passed to the correlators 33 provided in the  
35 remaining channels. One of these correlators is illustrated  
    in detail in Fig. 6.

1 As can be seen the signal received by the correlator 33  
of Fig. 6 from the reference channel is first split at a  
T-splitter 34 into two signals which are passed to res-  
pective inputs 35 and 36 of first and second linear multi-  
5 pliers 37, 38 with a dynamic range greater than 40 dB.

The signal coming from the respectively associated antenna  
19 is passed to a broad band  $90^\circ$  splitter 39 (bandwidth  
500 MHz on a 3 GHz system) which has two outputs 40 and 41.  
10 The signal at the output 40 has not been phase-shifted  
and is fed to a second input 42 of the first multiplier 37.  
The second output 41 has a phase shift of  $90^\circ$  relative to  
the input signal and is passed to the second input 43 of  
the second linear multiplier 38. The signals received at  
15 the inputs 35 and 42 have the general form

$$V_{35} = \tilde{V}_{18} \cos(\omega t - \varphi_{18}) \quad (G)$$

$$V_{42} = \tilde{V}_{19} \cos(\omega t - \varphi_{19}) \quad (H)$$

20

and the signals received at the inputs 36 and 43 have the  
general form

$$V_{36} = \tilde{V}_{18} \cos(\omega t - \varphi_{18}) \quad (I)$$

25

$$V_{43} = \tilde{V}_{19} \cos(\omega t - \varphi_{19} + \frac{\pi}{2}) \quad (J)$$

The signals appearing at the outputs 44 and 45 of the first  
and second linear multipliers 37 and 38 respectively have  
30 the general form:

$$S_{44} = \overline{V_{18} V_{19}} \cos(\varphi_{18} - \varphi_{19}) \quad (K)$$

$$S_{45} = \overline{V_{18} V_{19}} \sin(\varphi_{18} - \varphi_{19}) \quad (L)$$

35

The bar denotes the time average of the product of the  
signals  $\tilde{V}_{18}$  and  $\tilde{V}_{19}$ .

1 These signals are now integrated by converting them via  
 respective voltage to frequency convertors 46, 47 into  
 frequency signals which are then integrated in 16 bit  
 counters 48, 49 and fed onto a 16 bit databus via respect-  
 5 ive buffer stores 50, 51. A cycle of 8 such 20 msec samples  
 labelled A - H is given in the example shown in Fig. 7.  
 Each sample represents the correlated power received during  
 the sample period. The sign of the correlation is reversed for  
 each alternate sample by the phase switch 32, shown in  
 10 Figs. 5 and 6. The samples are combined by subtracting in  
 pairs thus (A-B), (C-D), (E-F), (G-H) for the example  
 given. This procedure effectively corrects for zero drifts  
 in the multipliers 37, 38 and the voltage to frequency  
 convertors 46, 47. The cosine correlation is thus formed  
 15 using elements 37, 46, 48 and 50 to form  $A_C, B_C, C_C, D_C,$   
 $E_C, F_C, G_C, H_C$  and elements 38, 47, 49 and 51 to form  $A_S,$   
 $B_S, C_S, D_S, E_S, F_S, G_S, H_S$ . These are fed by the 16 bit  
 databus 52 into a microprocessor which forms the com-  
 puting means and which is adapted to manipulate the first  
 20 and second values received from the buffer stores 50, 51  
 via the databus to calculate the relative phase and  
 amplitude at each antenna 19 relative to the reference  
 antenna 18. The relative phase and amplitude are calcul-  
 ated from the equations

25

$$\text{Cosine} = (A_C - B_C) + (C_C - D_C) + (E_C - F_C) + (G_C - H_C) = S_{44}, \quad (M)$$

$$\text{Sine} = (A_S - B_S) + (C_S - D_S) + (E_S - F_S) + (G_S - H_S) = S_{45}, \quad (N)$$

$$30 \text{ Then relative amplitude} = A = \sqrt{S_{44}^2 + S_{45}^2} \quad (O)$$

$$\text{relative phase} = \phi = \tan^{-1}(S_{45}/S_{44}) \quad (P)$$

These are computed for each correlated antenna pair.

35

The microprocessor is then programmed to calculate the  
 inverse Fourier transform from the relative phase and  
 amplitude values for each of the antennas in each of the

1 detection positions, and to produce the output as a matrix  
 of temperature values relating to adjacent volumes of said  
 dielectric medium. This matrix of temperature values can of  
 course be displayed in various ways, for example on the  
 5 screen of a monitor, possibly with different colours being  
 ascribed to different temperature values, and can also  
 be stored on a floppy disk for future reference or for  
 comparison with images made during successive examinations.

10 If, as previously-mentioned, it is found that the Fourier  
 transformation is not satisfactory because the array is  
 too close to the patient then it is possible to change  
 the relative phases calculated from the first and second  
 values by calculated or previously empirically derived  
 15 amounts and to improve the final result of the inverse trans-  
 formation.

Alternatively it is possible to calibrate the apparatus by  
 using a suitable body of dielectric with known high  
 20 temperature hot spots therein. The phase corrections  
 required to bring the measured distribution calculated by  
 the inverse Fourier transformation into agreement with the  
 actual temperature distribution can be recorded and  
 subsequently used to correct the relative phases calculated  
 25 when using the apparatus on a patient.

So far as carrying out the inverse Fourier transformation  
 is concerned the microprocessor is programmed to carry  
 out the following calculation:

30

$$f(l,m) = \iint g(x,y) \exp \left\{ 2\pi i (xl + ym) \right\} dx dy$$

In this expression  $l$  and  $m$  are the coordinates of a plane  
 close to the patient's body and the function  $f(s,m)$  is the  
 35 desired matrix of temperature values. For any given value of  
 $l$  and  $m$ ,  $f(l,m)$  gives the value of the temperature contri-  
 bution through the body projected onto this plane.  $x$  and  $y$

1 are the coordinates, expressed in wavelengths, of the plane  
 containing the antenna array. The origin of this coordinate  
 system, i.e.  $x = 0$ ,  $y = 0$  is preferably the reference  
 antenna and the origin of the plane close to the patient's  
 5 body, i.e.  $l=0$ ,  $m=0$  will be directly over this. The  
 function  $g(x,y)$  is complex and is the measured distribution  
 across the plane of the array. As an example for the  
 positions in Fig. 4 and taking antenna 18 as the reference  
 antenna, the measured value for  $g(x,y)$  will be  $g(x_{19}, y_{19}) =$   
 10  $= S_{44} + i S_{45}$ , where  $i = \sqrt{-1}$ .

When the patient is close to the array a Fourier transform  
 will not give adequate results. In this case close to  
 means less than the "far field" criterion given by  
 15  $z = 2 D^2/\lambda$  using the notation of Fig. 4. When  $z$  is only  
 slightly less than  $2 D^2/\lambda$  it will be possible to obtain  
 adequate results by putting in a phase correction to the  
 Fourier transform, but for smaller values of  $z$  the patient  
 is in the near field or Fresnel region. In this case the  
 20 microprocessor has to be programmed to carry out an inverse  
 Fresnel transform:

$$f(l,m) = \iint \exp \{i(xl + ym)\} g(x,y) \exp \{-i(a_x x^2 + a_y y^2)\} dx dy$$

25 In this equation  $a_x$  and  $a_y$  are parameters of the form:  
 $a_n = k_n \pi / r_n \lambda$  where  $r$  is the distance from the reference  
 antenna to the point in the  $l,m$  plane and  $r_n$  is the length  
 of the projection of  $r$  on the appropriate axis.  $k_n$  is  
 a parameter which will depend on the phase shifts along  
 30 the path  $a_n$ . The other symbols have their previous meanings.  
 This equation gives the transform for a two dimensional  
 picture of the temperature distribution, analogous to an  
 x-ray picture. It may be possible in practice to obtain  
 some estimate of the temperature distribution in the  $z$   
 35 direction, in which case the equation will need to be  
 modified appropriately.

1 One of the problems which can arise with thermographic  
apparatus is the problem of drift which can seriously  
affect the results. In order to overcome this drift, or at  
least to recognise it so that its effects may be taken  
5 into account, the apparatus may be provided with automatic  
gain control circuits which may be incorporated as a modification to the amplifiers 31. The present apparatus is  
provided with a continuous noise calibration system as a  
preferred alternative solution.

10

This system consists essentially of the noise diode 54  
illustrated in Fig. 5 which feeds a small noise signal into  
each of the antennas 18, 19. A circuit 55 is provided for  
modulating the output of the noise diode 54 with a square  
15 wave at a frequency  $F$ . The result of this is equivalent to  
turning the noise diode on and off at the frequency  $F$ ,  
the noise diode should however not actually be switched  
on and off as this would make it unstable.

20 The phase switch 32 present only in the electronic channel  
for the reference antenna serves to periodically change the  
phase of the signal in the reference channel by  $180^\circ$  and at  
a frequency  $2F$  which must be an integer multiple or vulgar  
fraction of  $F$ . The phase switch 32 is driven by a phase  
25 switch oscillator 32' which is linked synchronously to the  
noise switch oscillator 55' and the correlators 33, or  
alternatively (as shown in broken lines) to the noise switch  
oscillator 55' and the computer 53. This makes phase  
sensitive detection possible.

30

The effect of the noise calibration and of the phase switch  
is illustrated by an example shown in Fig. 7a over one  
sample period of 160 msec. It will be noted that this  
period of 160 msec is subdivided into eight portions each  
35 of length  $1/4F$ . As has been described earlier the circuit  
of Fig. 6 performs the correlated integral for each of the  
20 msec portions A - H and all eight values from each

1 multiplier are passed onto the databus. The microprocessor,  
in addition to computing the sine and cosine correlation  
terms using the formula  $(A-B)+(C-D)+(E-F)+(G-H)$  computes  
also the signal contributed by the noise diode 54 in  
5 Fig. 5 and which is demodulated using the formula  $(A-B)-$   
 $-(C-D)+(E-F)-(G-H)$ . This signal in a period of 160 msec  
has a significant measurement error due to noise but this  
can be reduced to a sufficiently accurate level by fitting  
a linear least squares solution to the noise diode cali-  
10 bration readings over a period of a minute or so. This  
signal can also be used to control the gain of the  
appropriate channel (automatic gain control AGC). In any  
case it is used in the software to provide a normalisation  
calibration factor so that changes in receiver gain can be  
15 compensated for in the analysis. As is shown in Fig. 7b  
this noise calibration vector must be subtracted from the  
outputs of the cosine and sine channel readings to get  
true amplitude and phase information.

20 When the reflection correction technique is used the noise  
diode is used to inject noise into the antenna lines so that  
this signal is radiated out into the dielectric media (at  
a very low level). This corrects for reflection losses at  
dielectric boundaries. The amplitude and phase of the  
25 reflected signal can be demodulated in the same way as  
the gain calibration signal, and has also to be subtracted  
from the sine and cosine channels to get the true readings.  
This noise reflection correction technique has been des-  
cribed in U.S. Patent 4,235,107 for single antenna measure-  
30 ments.

As mentioned earlier it will probably be necessary to  
take account of the state of polarisation of the signals  
received in each channel. Polarisation components can be  
35 defined and measured in many ways but all methods of  
determining them in a synthesis system rely on antenna  
elements at each antenna position that each respond to one



1 or both polarisation components. The signal from one element (e.g. 18 in Fig. 6) is then combined with the signal from either its "double" or its "complement" at the other antenna position (19, Fig. 6). The other pair then requires  
5 another set of electronics (Figs. 5+6). Full polarisation data is then obtained by either adding or subtracting the outputs of the two pairs either in hardware or in software before or after the transformation. The combination chosen is a design parameter. It may however also be  
10 possible to make a combined antenna that responds to more than one type of polarisation giving an output proportional to the total signal arriving at that point.

In order to provide good contact between the patient and  
15 the table it may be desirable to use a high dielectric constant cushion which may be capable of being moulded to the patient's body shape or a thin polythene envelope of warm water. In either case the intention is to avoid a patient air interface which would give rise to undesirable  
20 losses.

Finally, the following comments are made concerning the system parameters and components.

#### 25 a) Spatial Resolution

Resolution is used here in the sense of the full beam width given by  $\theta = \lambda/D$ . For an array at a distance  $z$  from the object being examined the resolution is then  $r = \lambda z/D$  cm,  
30 expressed in terms of distance in the patient (see Fig. 4).  $D/z$  can only have a value between 0.5 and 5 for practical reasons, with 4 being a reasonable value, but the wavelength can be reduced by a factor of 5.5, using a dielectric with  $\epsilon_r = 30$  between the patient and the array giving  
35 a corresponding improvement in resolution. This gives a resolution of

1

$$r = \frac{cz}{\nu D \sqrt{\epsilon_r}} \quad (Q)$$

5 where  $c$  is the velocity of light, and  $\nu$  is the observing frequency =  $\omega/2\pi$ .

$r$  is given in Table 1 (page 33) for several frequencies with  $D/z = 4$ . The use of a dielectric has two further advantages: (i) the small antennae can be mounted on a rotating piece of similar dielectric enabling strip line techniques to be used in their construction, and perhaps also in the local oscillator supply, and signal amplifier and mixer chains; (ii) suitable shaping of the material to give good contact to the patient's body-shape will considerably improve the matching of the body to the antennae as there is then no longer a skin to air interface (high to low  $\epsilon_r$ ).

20 The resolution given above, which is of the order of 0.5 to 2.5 cm for  $3 \text{ GHz} > \nu > 1 \text{ GHz}$ , is not the error in the determination of the position of the hotter region, as beam-fitting techniques can give the position of small sources to within 0.1 beam widths even when the signal-to-noise ratio of the signal is only 5:1. In this context beam fitting means numerical techniques which give a best fit of a diffraction pattern to a small part of the matrix of temperature values.

### 30 b) Operating Frequency

The choice of frequency must consider four factors affecting the visibility of a hot region of fixed size inside the body:

35

- 1 (i) the radiation's penetration (1/e) depth in tissue increases with wavelength, the absorption coefficient is given approximately by the following empirically derived formula:

5

$$\alpha = 2.2 \times 10^{-5} \sqrt{\nu} + 0.1 \quad (R)$$

over the frequency range 100 MHz to 3 GHz;

- 10 (ii) the hot spot's optical depth increases with frequency giving an increase in effective temperature rise above the surroundings;

15

Optical depth ( $\tau$ ) is a measure of the absorption of radiation of a particular wavelength as it passes through a lossy medium. For a piece of this medium, the radiation intensity emerging is  $e^{-\tau}$  times the incident radiation intensity. For black-body radiation from the lossy medium, which we consider here, the intensity arising from it will be  $(1 - e^{-\tau})T$  where T is the temperature of the medium;

20

- (iii) when the hot-spot's diameter is less than the diameter of the antenna arrays synthesised beam at that point the hot-spot's measured temperature is reduced by the ratio of the beam diameter to hot-spot diameter;

25

- 30 (iv) the reflections at interfaces inside the body between materials of different permittivities cause losses.

Quantifying effects i, ii and iii, and using equations (A) and (B), lead to a relation for the change in temperature seen at the skin's surface

35

$$\Delta T_e = \Delta T \epsilon_r \left( \frac{D k \sqrt{\nu}}{zc} \right)^2 \left[ 1 - \exp \left\{ - (2.2 \times 10^{-5} \sqrt{\nu} + 0.1) k \right\} \right] \cdot \exp \left\{ - (2.2 \times 10^{-5} \sqrt{\nu} + 0.1) b \right\} \quad (S)$$

where:  $\Delta T$  is the hot spot's actual temperature rise above its surroundings, = 1.5 K in this example,

$k$  is the diameter of the hot spot assumed to be a uniform disc of thickness  $k$ , = 1 cm in this example,

$b$  is the depth of the hot spot below the body surface.

The value of  $\Delta T_e$  for several parameters is shown in Fig. 8a. Fat has been ignored here because its absorption coefficient is much lower than above, but shows the same variation with frequency. The effect of mismatches at boundaries due to effect (iv) above can be seen in Fig. 8b which gives the temperature measured just outside the skin for a 1.5 cm thick layer of hotter material seen through 2 mm of skin and layers of muscle and fat. This is based on the following relation which has been mathematically derived from the information given in the paper by J. Edrich in "The Journal Of Microwave Power" vol. 14, No. 2, 1979, pages 95 to 104:

$$\Delta T_e = \Delta T \left[ 1 - \exp \left( - \frac{z_t}{d_t} \right) \right] \cdot \exp \left( - \frac{z_f}{d_f} \right) \cdot e_{fs} \cdot \exp \left( - \frac{z_s}{d_s} \right) \cdot e_{so} \cdot \exp \left( - \frac{z_m}{d_m} \right) \cdot e_{mf} \quad (T)$$

where:  $z_t, z_f, z_s, z_m$  are the thicknesses of the layers of tumour, fat, skin and muscle respectively,  
 $d_t, d_f, d_s, d_m$  are the  $1/e$  penetration depths of the materials at a given frequency,  
 $e_{fs}, e_{so}, e_{mf}$  are the emissivities from fat to skin, skin to outside the body, and muscle to fat.

1 Consideration of Fig. 8 shows that for small hot spots  
 higher frequencies are better, until either  $\Delta T_e$  levels  
 off or equation (R) breaks down ( $> 3$  GHz), whilst for  
 larger hot areas the lower frequencies are better. There-  
 5 fore the choice of operating frequency, within the range  
 900 to 3000 MHz, must depend on the particular application  
 being considered.

Closely related to the operating frequency is the usable  
 10 bandwidth. This must be sufficiently small for the phase  
 difference between signals at the antennae with the largest  
 separation,  $A_1$  and  $A_2$  in Fig. 4, not to change by more  
 than  $90^\circ$  across the band for points at the edge of the  
 picture area. Taking the picture area to be the same as  
 15 the antenna array area and directly over it, we come to  
 the relation:

$$\Delta \nu < \frac{\nu^2}{4c} \frac{xz}{(x-z)} \quad (U)$$

20 where  $\Delta \nu$  is the bandwidth and  $x = z^2 + D^2$ . This is a very  
 weak constraint, varying from 11 % to 40 % of the operating  
 frequency over the range 915 MHz to 3000 MHz, and the  
 bandwidths given on page 33 are less than this.

## 25 c) The Antenna Array

In the far-field case, the number of antennas is determined  
 by the minimum separation between antennas ( $\Delta D$ ) for the  
 lowest Fourier component to have a radius larger than the  
 30 picture size. In our case, but with far-field approximat-  
 ions, the restriction is that  $\lambda/\Delta D > \phi$  where  $\phi$  is  
 the angle subtended by the examined area at one element  
 of the array, with  $\phi = \tan^{-1} (D/z)$ . Expressed in terms  
 of frequency we have:

$$\Delta D < \frac{c}{\nu} \left[ \epsilon_r \tan^{-1} \left( \frac{D}{z} \right) \right]^{-1} \quad (V)$$

and the number of antennae.

1

$$n = \frac{D}{\Delta D} + 1 \quad (W)$$

For  $D/z = 4$ ,  $\Delta D$  is of the order of one or two centimetres  
 5 so dipoles, with suitable balancing networks, are ideal  
 as the individual antennas. The patient is in the near  
 field of the array so it is not possible to reduce the  
 number of antennas by the usual procedure of having un-  
 equally spaced antennas separated by multiples of  $\Delta D$  and  
 10 reconstructing the whole aperture using combinations of  
 these as used in radio astronomy. Although this increases  
 the array costs it is offset to some extent by the fact  
 that the path compensators and phase rotators used in  
 astronomy are unnecessary and the data rate is not too  
 15 high for a small computer handle.

#### d) Preamplifiers

Many preamplifiers are necessary as all antenna pairs are  
 20 present so they must be inexpensive. Since the array looks  
 at a 300 K patient this is the lowest possible overall  
 system noise temperature with a perfect amplifier. With the  
 calibration methods discussed in the specification direct-  
 ional coupler, circulator and filter are needed between  
 25 the dipoles and preamplifiers giving a small degradation  
 in system noise temperature. A simplified sketch of the  
 system is shown in Fig. 5. For our calculations we have  
 assumed a system noise temperature of 600 K = 300 K patient  
 + 300 K amplifier since such amplifiers are readily  
 30 available at low cost.

#### e) The Correlators and Continuous Calibration

The amplitude (A) and phase ( $\phi$ ) are measured as sine (S)  
 35 and cosine (C) terms, where  $A^2 = S^2 + C^2$  and  $\phi = \tan^{-1}(S/C)$ ,  
 by cross correlation of the signal from the central receiver  
 with that of a receiver at the appropriate spacing for each

1 of the spacings. In the system proposed the bandwidths are  
much larger than those used in astronomical synthesis so  
the correlators must be carefully designed so that the  
relative phase between inputs is less than a few degrees  
5 across the whole bandwidth (200 MHz for the 1.5 GHz system).

In addition they need at least a 40 dB dynamic range  
(1 mK to 10 K). This specification can be achieved by the  
use of linear multipliers and by 180° broad band phase  
10 switching in one of the I.F. inputs (see Fig. 6). When  
synchronous demodulation is applied this removes the 5 %  
or so square law response inherent in the multipliers at  
higher signal levels and simultaneously corrects for zero  
drift in the multipliers and following voltage-to-frequency  
15 convertors. These convertors, in conjunction with buffered  
counters, act as integrators. Gain calibration for each  
channel is derived by a modulated noise source switched  
at half the phase inversion rate and which is weakly  
(-25 dB) but coherently coupled into the input circuit of  
20 each preamplifier, see Fig. 5. Software demodulation makes  
it possible to carry out real time correction, during the  
analysis, for system drifts of phase and amplitude and  
provides engineering monitoring of possible system mal-  
functions.

25

#### f) Inverse Transform and Image Matrix

During the rotation of the array we need to collect  $D^{\sim}/\Delta D$   
30 samples each of N-1 amplitudes and phases to match the mini-  
mum separation limit calculated in section c. For the  
1.5 GHz system this gives 57 samples each of 18 amplitudes  
and phases. We have chosen 64 samples each lasting  
160 milliseconds in the software demodulation example previ-  
35 ously described. As the array rotates the amplitude and  
phase information is interpolated into a matrix of values  
in the aperture plane and the continuous calibration

1 corrections previously described are computed and applied  
at the end of the scan directly into the aperture plane  
array. This array is then transformed to form the image.  
This transform may well require special purpose hardware  
5 to take full advantage of the short 10 second exposure time  
required by the receivers.

The image matrix requires a minimum of two points per  
resolution element which for the 1.5 GHz system is 0.91 cm  
10 (Table 1, page 27) giving a grid point separation of 0.45 cm.  
A 50 cm square image then needs a matrix of 111 x 111  
points. If each element has 2 byte accuracy (1 in 65536) we  
can provide a temperature range of 10 K relative to some  
conveniently chosen zero for the image with an accuracy of  
15 0.15 mK due to round-off. Such an image would require 25 K  
bytes including identification and calibration information  
and sixteen such images could be stored on a single  
0.5 Megabyte floppy disk including patient case history  
details. This would give a convenient cheap filing  
20 method for hospital use, allowing later off-line inspection  
by a consultant and eventual transfer to mass storage for  
large sample statistical analysis.

The image matrix could well have the format described in  
25 the article "NOD 2 A general system of analysis for  
radioastronomy" by C.G.T. Haslam in Astronomy and Astro-  
physics Supplement vol. 15 p.p. 333-335, 1974 which is  
very efficient in computer time when used with two-  
dimensional regridding and interpolation routines. These  
30 operations will be essential for comparing pictures taken  
on different days, since the patient will not lie in  
exactly the same place, so superposition, shifts of scale,  
translation and rotation will be required before detailed  
comparison can be made. Suitable software and necessary  
35 algorithms, including techniques for combining over-  
lapping images, have been described, in an astronomical  
context, in the article "NOD 2 A Generalised System of  
Data Analysis for Astronomy which has Application to Image



1 Processing" by C.G.T. Haslam, N.C. Haslam and D.T. Emerson,  
Technischer Bericht 56 des Max-Planck-Instituts für  
Radioastronomie Bonn, 1980.

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Table 1**System Parameters.**

Maximum antenna separation 50 cms.

Distance from patient to antenna array 12.5 cms.

Dielectric constant of table 30

System noise temperature 300 K + 300 K from patient.

Frequency	915	1500	2450	3000	MHz
Resolution	1.49	0.91	0.55	0.45	cms
Number of antennae in array	12	19	30	37	
Bandwidth	100	200	400	500	MHz
R.m.s. noise in 10 second exposure	5.47	3.07	1.73	1.39	milli Kelvin
Time taken to see 1.5 K warmer layer below 2 mm skin, 1 cm fat and 4 cm muscle	6	9	57	222	seconds

- 1 1. Thermographic apparatus for measuring the temperature  
distribution in a substantially dielectric medium by  
detecting the microwave energy emerging from the medium,  
the apparatus comprising an array (13; 13') of micro-  
5 wave antennas (18, 19) positioned adjacent the dielectric  
medium and electronic means (Figs. 5 + 6) for processing  
the broad-band signals induced in the microwave antennas  
(18, 19) in a plurality of detecting positions (21) to  
10 determine the temperature prevailing in a volume element  
of the dielectric medium, characterised in that the  
electronic means (Figs. 5 + 6) comprises means (25, 26,  
27, 28, 29, 29', 30, 31, 33) for processing the signals  
induced in each of the antennas (18, 19) in each of the  
15 detecting positions (21) and for correlating each signal  
with the signal from at least one of said antennas (18)  
in a fixed reference location to produce, for the or each  
detecting position (21) of each antenna (19), first and  
second values containing information relating to the  
20 amplitude and phase of the signal received in that de-  
tecting position relative to the amplitude and phase  
of the signal received by at least one of the reference  
antennas (18); and computing means (53) for forming an  
inverse transformation of the first and second values  
associated with each antenna of the array, the result of  
25 said inverse transformation being a matrix of temperat-  
ure values relating to adjacent volumes of said dielec-  
tric medium.
2. Thermographic apparatus in accordance with claim 1 and  
30 characterised by the provision of phase adjustment means  
for adjusting the phase of each signal induced in each  
of the antennas (19) in the or each detecting position,  
relative to the phase of the signal received by at least  
one of the reference antennas (18), prior to effecting  
35 the inverse transformation, whereby to compensate for  
near field effects.

- 1 3. Thermographic apparatus in accordance with either of  
the preceding claims, characterised in that an inert  
dielectric layer (11) of high permittivity, preferably  
greater than 10, and of low loss, preferably better than  
5 5 %, is placed between said array (13, 13') and said  
medium.
4. Thermographic apparatus in accordance with claim 3,  
characterised in that said dielectric layer (11) is  
10 greater than 5 cms thick.
5. Thermographic apparatus in accordance with any one of  
the preceding claims, characterised in that each of the  
antennas (18, 19) is replaced by two or more antenna  
15 elements whereby separate polarisation components can  
be measured either prior to or after the inverse  
transformation, and, optionally, in that means are pro-  
vided for compensating for polarisation differences.
- 20 6. Thermographic apparatus in accordance with any one of  
the preceding claims, characterised in that means (29,  
29'; 30, 31) are provided for varying the centre  
frequency and/or the bandwidth at which the measurement  
of the temperature distribution is carried out.
- 25 7. Thermographic apparatus in accordance with claim 6,  
characterised in that said electronic means (Figs. 5 + 6)  
comprises a local oscillator (30) and a mixer (29) for  
generating a difference frequency, with said local  
30 oscillator frequency being variable whereby to vary said  
centre frequency.
8. Thermographic apparatus in accordance with claim 7,  
characterised in that amplifier (31) and/or filter means  
35 is provided for varying said bandwidth.

- 1 9. Thermographic apparatus in accordance with claim 6  
wherein said electronic means does not include a mixer  
and/or local oscillator, characterised in that the centre  
frequency and bandwidth are determined by an amplifier.

5

10. Thermographic apparatus in accordance with any one of  
the preceding claims, characterised in that said array  
(13') of microwave antennas (18, 19) is spatially fixed,  
whereby each antenna (18, 19) has one detecting  
10 position; and in that switching means is provided for  
connecting said antennas groupwise in turn to said elec-  
tronic means (Figs. 5 + 6).

11. Thermographic apparatus in accordance with any one of the  
15 preceding claims 1 to 9, characterised in that means  
(16) is provided for moving said array (13) so that each  
antenna (19) other than said reference antenna (18) or  
antennas (18') adopts or moves through a plurality of  
detecting positions.

20

12. Thermographic apparatus in accordance with claim 11,  
characterised in that said movable array comprises one  
or more linear arrays (13).
- 25 13. Thermographic apparatus in accordance with claim 11,  
characterised in that said movable array comprises a  
plurality of non-linearly spaced antennas.

14. Thermographic apparatus in accordance with any one of  
30 the preceding claims, characterised in that intermediate  
storage means is provided for temporarily storing  
said signals from each of said antennas (18, 19) in the  
or each detecting position.

- 35 15. Thermographic apparatus in accordance with any one of  
the preceding claims, characterised in that said elec-  
tronic means (Figs. 5 + 6) includes a noise calibration

- 1 system (54, 55, 55') with a noise generating source (54),  
such as a noise diode, adapted to feed noise to each of  
said antennas (18, 19); and in that said calibration  
system (54, 55, 55') is optionally adapted to calibrate  
5 the apparatus to take account of reflections at the  
surface of the medium and/or at boundaries within the  
medium.
16. Thermographic apparatus in accordance with any one of the  
10 preceding claims, characterised in that said electronic  
means (Figs. 5 + 6) further comprises means for con-  
trolling the gain of the said electronic means automatic-  
ally, said automatic gain control being optionally con-  
trolled in dependence on a signal derived from the said  
15 noise calibration system.
17. Thermographic apparatus in accordance with any one of  
the preceding claims characterised in that said elec-  
tronic means (Figs. 5 + 6) comprises first multiplier  
20 means (37) for forming the products of the signal from  
the reference antenna (18) with the signal induced in  
each of the remaining antennas (19) in the or each  
detecting position to yield said first values; phase  
shifting means (39) for shifting the phase of the  
25 signal induced in each of the remaining antennas in  
the or each detecting position by  $90^{\circ}$  to produce phase  
shifted signals; and second multiplying means (38) for  
forming the products of the signal from the reference  
antenna (18) with each of the phase shifted signals to  
30 yield said second values.
18. Thermographic apparatus in accordance with claims 15 and  
17 characterised in that said electronic means (Figs. 5  
+ 6) further comprises means (32) for periodically  
35 changing the phase of the signal from the reference  
antenna by  $180^{\circ}$  with the change taking place at a first  
frequency (2F) and optionally further means (55) for

1     periodically switching the output of said noise source  
     (54) between a selected value and zero at a second  
     frequency (F) equivalent to said first frequency (2F)  
     divided by or multiplied by an integer, and means for  
5     periodically sampling said first and second values at  
     intervals of time (1/4F) equivalent to one half of the  
     reciprocal of said first frequency ; and in that said  
     computing means (53) is adapted to process said sampled  
     first and second values to derive calibration information  
10     and signal information.

19. Thermographic apparatus in accordance with any one of  
     the preceding claims characterised in that said antennas  
     comprise dipole antennas (18, 19) or antennas formed as  
15     printed circuits on dielectric material.

20. The combination of thermographic apparatus in accordance  
     with any one of the preceding claims with hyperthermia  
     apparatus characterised by means for operating the  
20     thermographic apparatus and the hyperthermia apparatus  
     in alternative time periods, optionally using the same  
     antennas, whereby to detect the temperature distribution  
     produced by the hyperthermia apparatus and, optionally,  
     by means for controlling the hyperthermia apparatus  
25     using the signals from the thermographic apparatus.

21. The combination of thermographic apparatus in accordance  
     with any one of the preceding claims with a narrow band  
     noise source located on the side of said dielectric  
30     medium opposite to said thermographic apparatus, wherein  
     said narrow band noise source is blinked and detected  
     synchronously by said computing means.

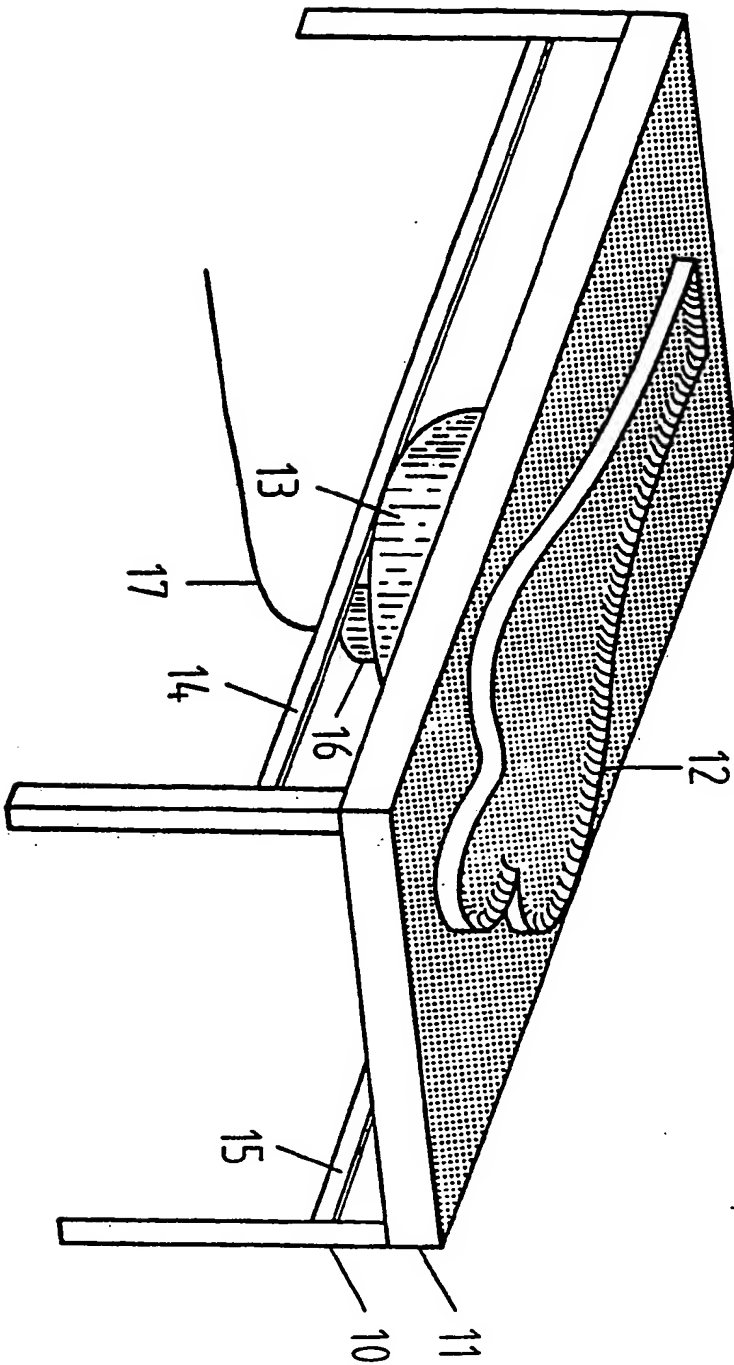


FIG. 1



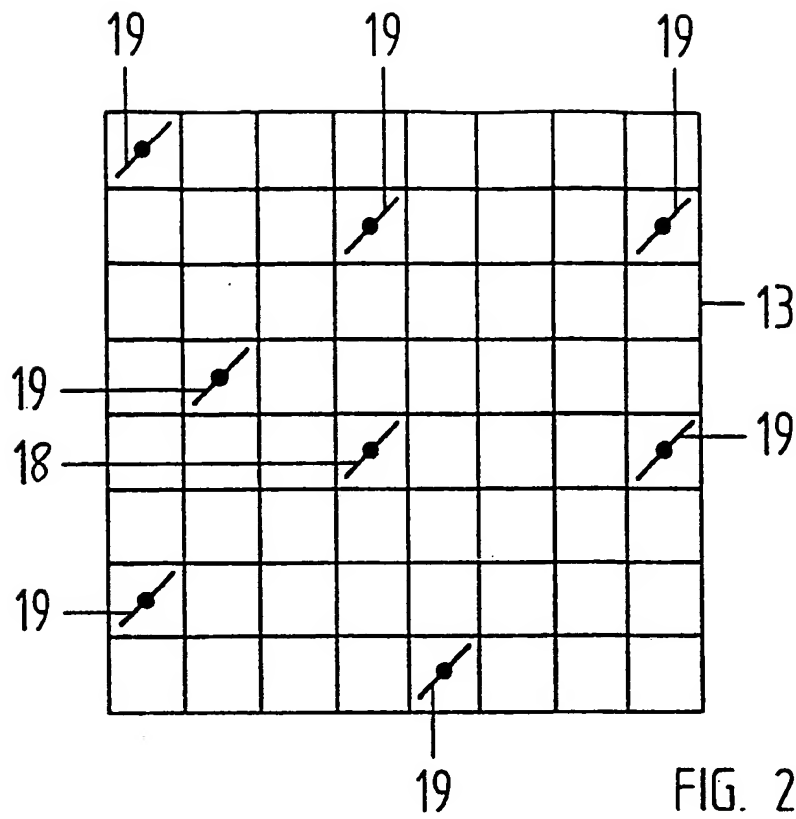


FIG. 2

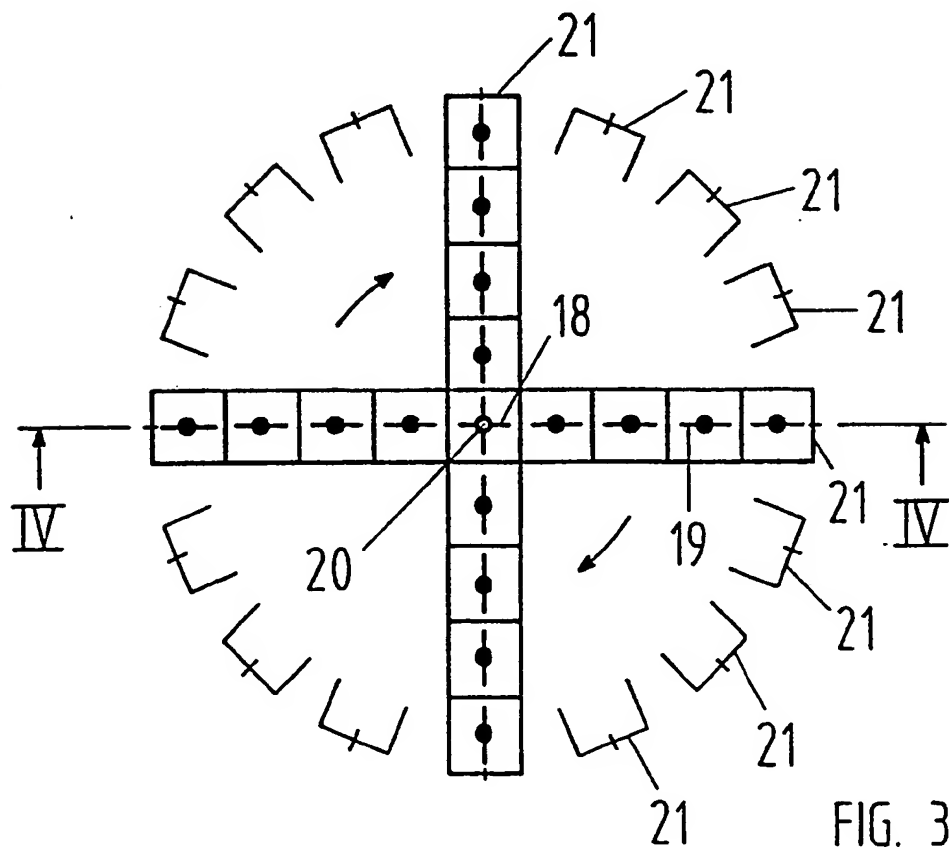


FIG. 3a

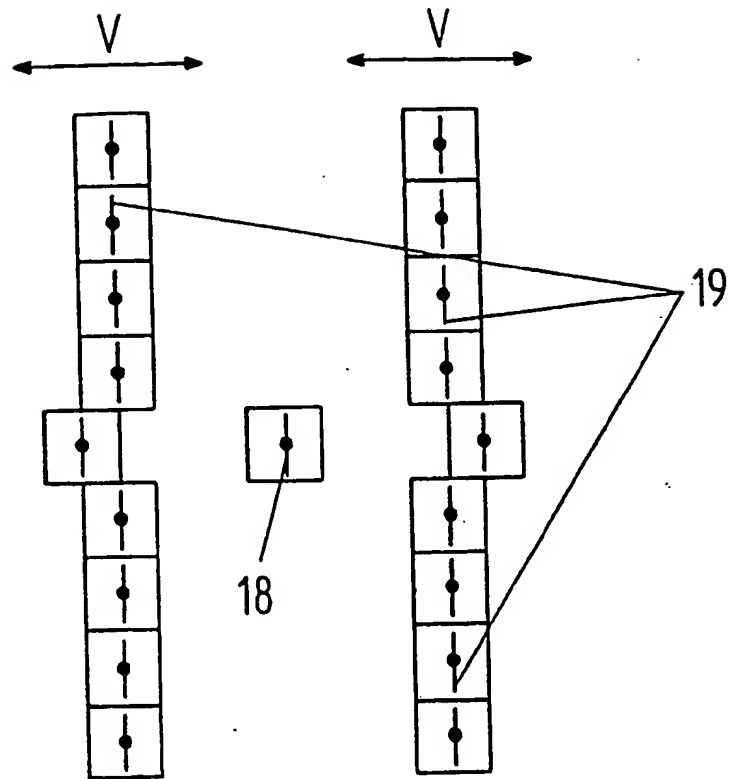


FIG. 3(b)

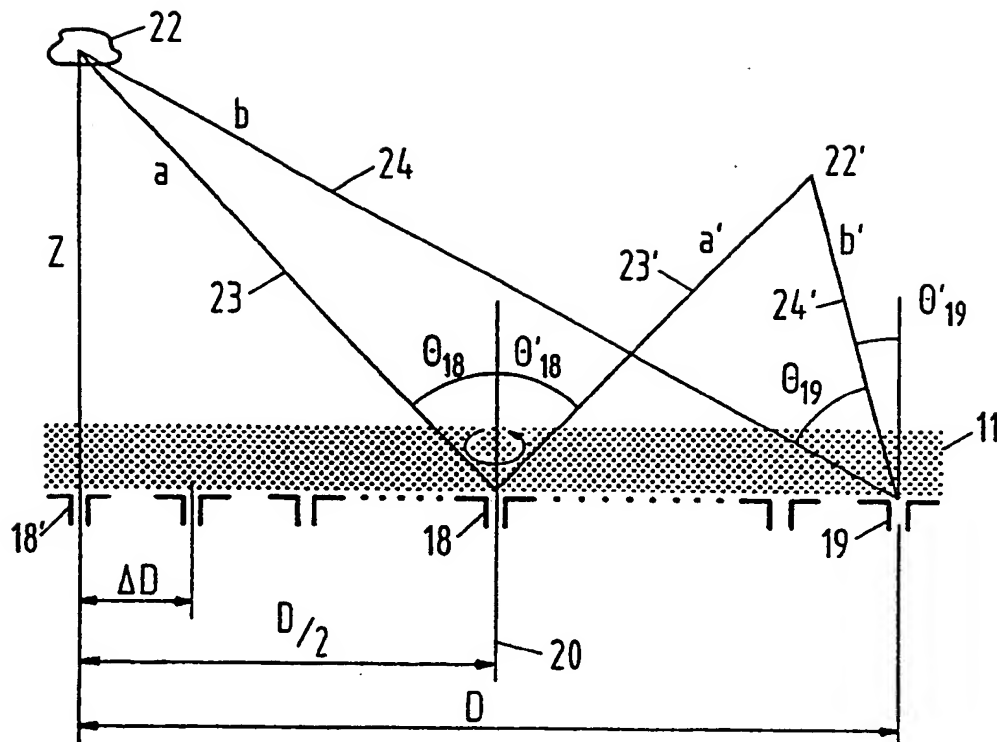


FIG. 4



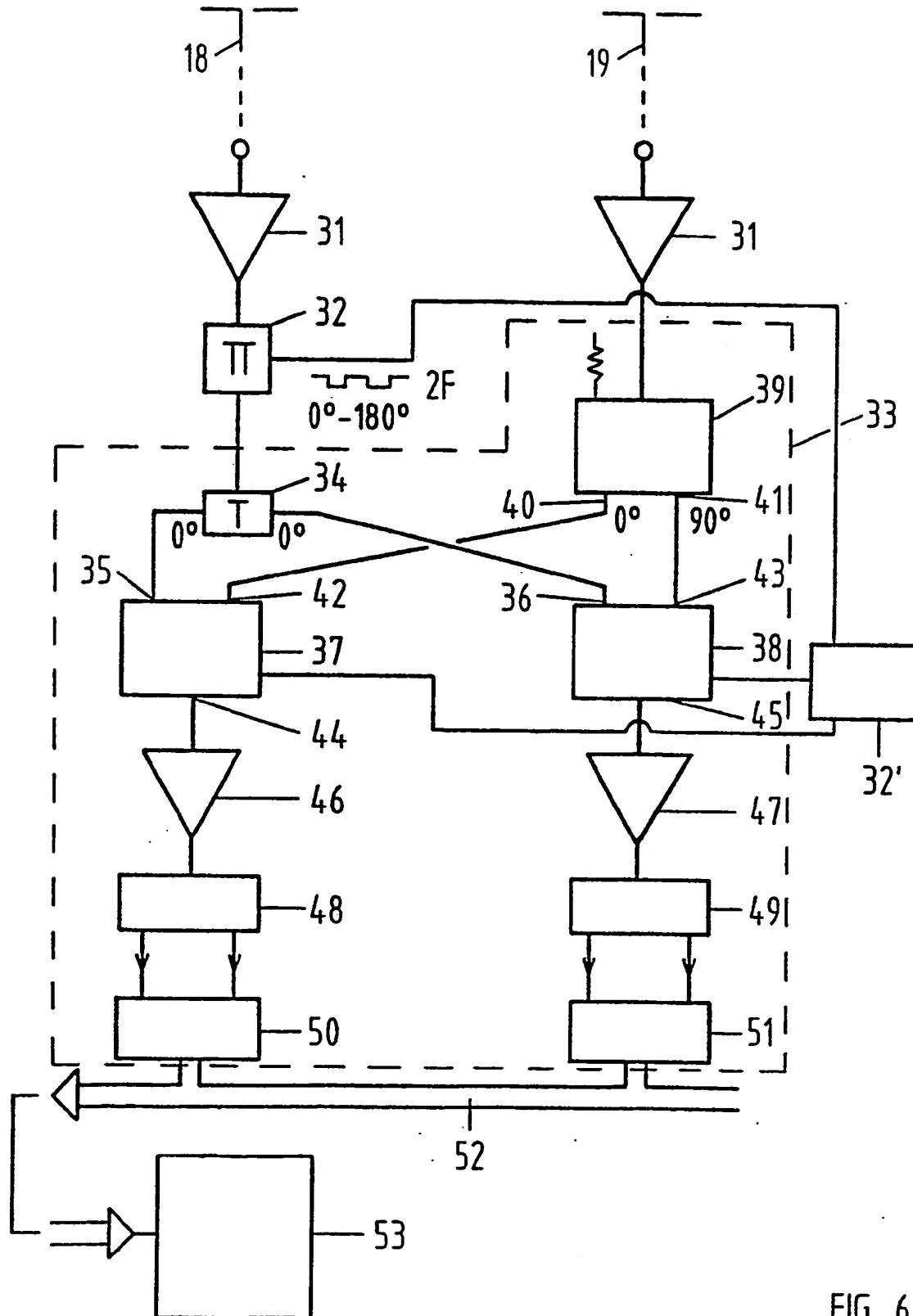


FIG. 6

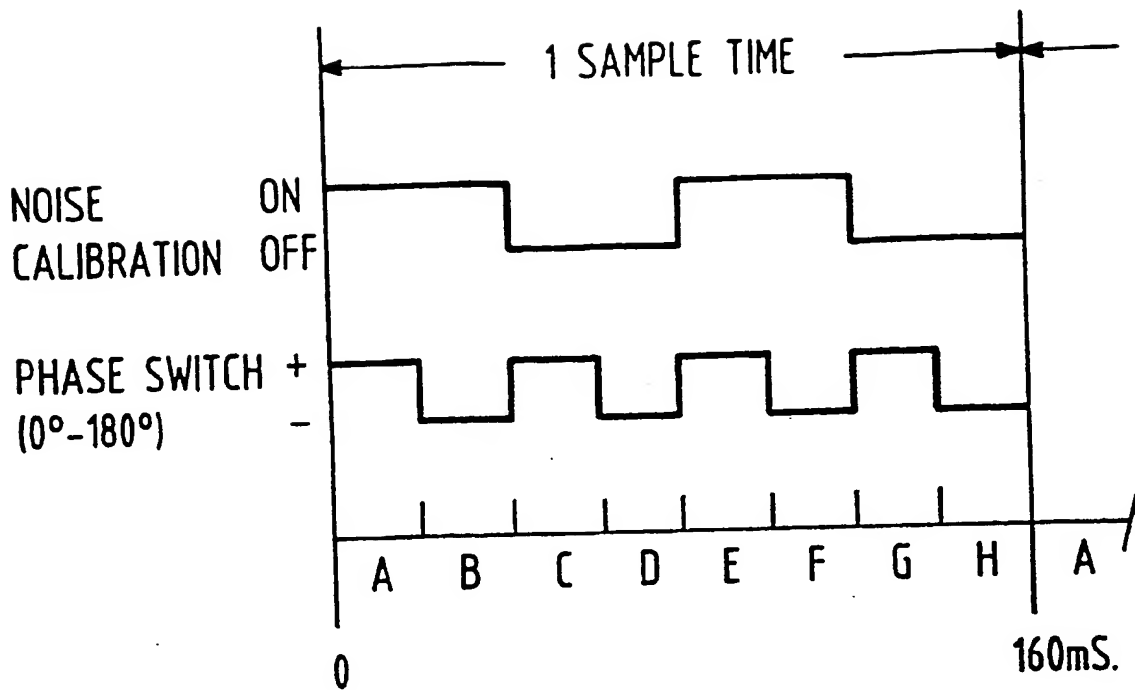


FIG. 7a

TRUE AMPLITUDE AND  
PHASE TRACK

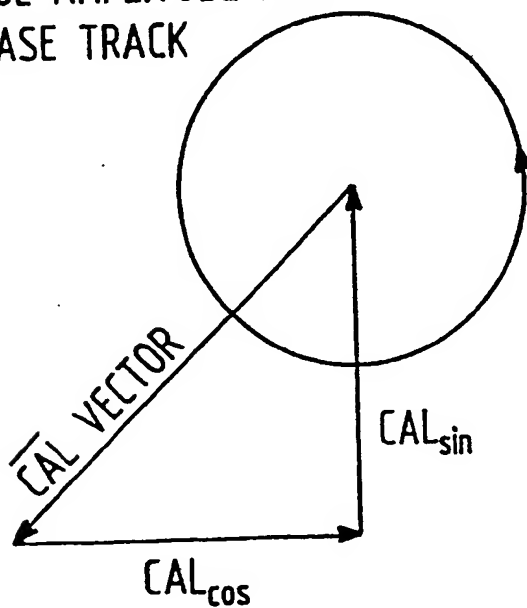


FIG. 7b

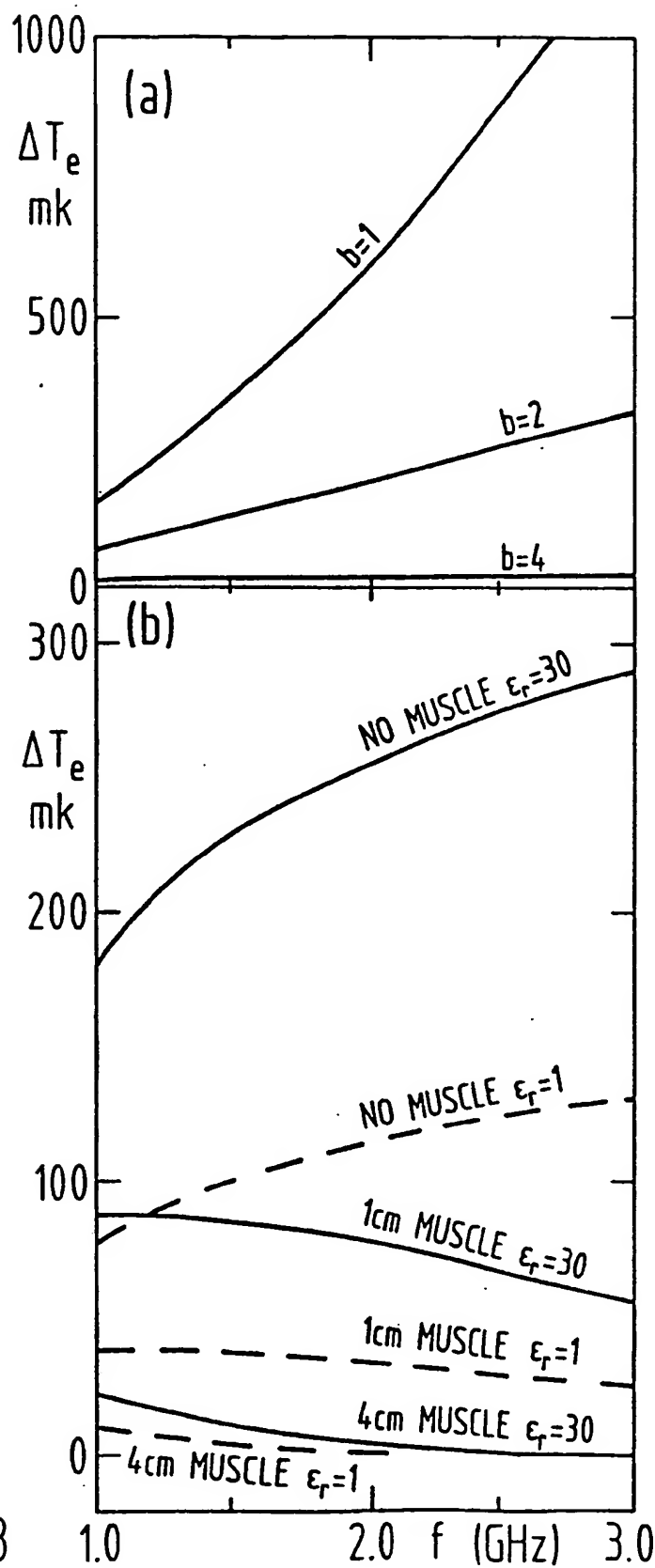


FIG. 8



DOCUMENTS CONSIDERED TO BE RELEVANT			
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl.4)
Y	GB-A-2 052 909 (INSTRUMENTARIUM CY.) * Page 2, lines 10-40; figure 1 *	1	G 01 R 29/08
A		3	
Y	--- CONFERENCE PROCEEDINGS, 12th EUROPEAN MICROWAVE CONFERENCE, 13th-17th September 1982, pages 553-558, Helsinki, FI; A. MAMOUNI et al.: "Introduction to correlation microwave thermography" * Page 554, figure 1; paragraph: "Principle of the correlation microwave thermography" *	1	
Y	--- US-A-4 178 100 (ROBERT A. FROSCH) * Abstract; figure 2 *	1	TECHNICAL FIELDS SEARCHED (Int. Cl.4)
A		6, 12, 15, 16	G 01 R A 61 B A 61 N
A	--- US-A-4 416 552 (ROBERT A. HESSEMER, LLOYD J. PERPER) * Figures 12-16; claims 1, 2, 4-7, 14-15, 18 *	5-9	
	--- -/-		
The present search report has been drawn up for all claims			
Place of search THE HAGUE		Date of completion of the search 10-12-1984	Examiner KAUFFMANN J.
<p>CATEGORY OF CITED DOCUMENTS</p> <p>X : particularly relevant if taken alone Y : particularly relevant if combined with another document of the same category A : technological background O : non-written disclosure P : intermediate document</p> <p>T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons</p> <p>&amp; : member of the same patent family, corresponding document</p>			





DOCUMENTS CONSIDERED TO BE RELEVANT			
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl. 4)
A	1982 IEEE MTT-S INTERNATIONAL MICROWAVE SYMPOSIUM DIGEST, 15th-17th June 1982, session paper V2, pages 438-440, Dallas, USA; F. STERZER et al.: "A self-balancing microwave radiometer for non-invasively measuring the temperature of subcutaneous tissues during localised hyperthermia treatments of cancer" - Figure 2 *	19,20	
A	IEEE MTT-S INTERNATIONAL MICROWAVE SYMPOSIUM DIGEST 1983, 31st May - 3rd June, Session F4, pages 186-188, Boston, USA; Y. LEROY et al.: "Present results and trends in microwave thermography"		TECHNICAL FIELDS SEARCHED (Int. Cl. 4)
The present search report has been drawn up for all claims			
Place of search THE HAGUE		Date of completion of the search 10-12-1984	Examiner KAUFFMANN J.
CATEGORY OF CITED DOCUMENTS		T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document	
X	particularly relevant if taken alone		
Y	particularly relevant if combined with another document of the same category		
A	technological background		
O	non-written disclosure		
P	intermediate document		